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Effects of Pulsed Laser Systems on Stapes Footplate

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Background and Objective: **The aim of the present study was to investigate the tissue ablation capacity of various pulsed lasers at the stapes footplate.**
Study Design/Materials and Methods: **Isolated human stapes and bovine compact-bone platelets were used to determine the effective laser parameters and appropriate application technique for achieving a perforation measuring 500–600 μm in diameter. Of interest were also the shape and quality of the perforations, the reproducibility of the perforation effect, and the thermically altered marginal zones occurring at the footplate. Three pulsed laser systems were used: excimer, holmium:YAG (Ho:YAG), and erbium:YSGG (Er:YSGG) lasers.**
Results: **The tissue-ablating effect of pulsed laser systems permits a precise and controlled management of the stapes footplate through low and readily reproducible ablation rates. The extent of thermic side effects at the footplate is lower in comparison to the purely thermically acting cw and superpulse laser systems. The Er:YSGG laser exhibits the highest ablation rate at the stapes and is thus the most effective laser for perforation of the stapes footplate. Though somewhat less effective, the Ho:YAG laser also appears to be suitable for stapedotomy. On the other hand, we do not consider the applied excimer laser (308 nm) to be particularly appropriate at the stapes because of its low ablation rates.**
Conclusion: **Thus, the erbium laser could represent an alternative to the argon, KTP 532, and CO_2 lasers, already clinically successful in stapes surgery. However, further studies are neccessary to examine the transmission of thermic energy into the vestibule and the acoustic stress to the inner ear during laser stapedotomy, to be able to make a definitive statement about the safest and most effective laser system for stapes surgery. Lasers Surg. Med. 21:341–350, 1997.**

Key words: Er:YSGG laser; excimer laser; Ho:YAG laser; in vitro; laser stapedotomy, laser tissue effect

INTRODUCTION

The idea of using the laser as a non-contact instrument in stapes surgery is based on the desire to reduce the complication rate of this inter-

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vention by further optimization of the operation technique. The complications occuring in stapes surgery (varying degrees of sensorineural hearing loss extending to deafness, etc.) are attributed to various causes, above all to the inadequacy of the manual technique with damage to Corti's organ.

The aim of laser stapedotomy is to achieve stapes footplate perforation in such a way as to ensure that the inner ear will be endangered as little as possible and that there will be no damage to the remaining middle ear structures. In our study, perforation of the footplate was therefore performed in a non-contact manner with the laser beam.

The continuous-wave lasers have been examined as to their suitability for stapedotomy in previous studies [2–5,15]. This study aims at investigating various pulsed laser systems which, among other things, have yielded good results in osteotomy research [1,8,12] and thus also promise to be suitable for management of the auditory ossicles.

The effectiveness and safety of the thermically acting continuous-wave (cw) lasers (argon, KTP 532, and CO_2 lasers) hitherto applied in clinical practice are still discussed controversially [3,6,9]. Novel types of pulsed laser systems (excimer, holmium:YAG, and erbium:YAG lasers) that can act oligothermically may prove to be safer and more efficient [2,7,10,11,13,14]. The results of these studies are checked and reassessed with adequate experimental and analytical methodes.

In a range of short pulse durations and high power densities (about 10^8 W/cm^2), the occurrence of so-called non-linear processes lead to a change in the effect of the laser on tissue.

The resultant phenomena differ markedly from the purely thermic effects of the laser application and lead to changes in the ablation mechanism and marginal effects. This process, which is designated as photoablation, takes place at energy densities of about 0.1–10 J/cm^2 and laser pulse durations in the nano- and microsecond range. With this process, the exposure times and thus the duration of the temperature increase are so short that heat conduction is negligible. Tissue ablation is thus achieved with low heating of adjacent structures and thus only small thermically influenced zones around the ablated tissue. The ablation process can be evaluated by determining the ablation rate, which indicates thickness of the tissue layer ablated per laser pulse.

In describing the pulsed laser systems, it is essential for an effect analysis to specify the wavelength λ (μm), the pulse energy Q (J), the energy density H (J/cm^2), the pulse duration (full width at half maximum (FWHM)) t (μs), and the pulse repetition rate (Hz).

For the wavelength of the CO_2 laser (λ = 10.6 μm), the absorption coefficient for compact bone and the human stapes footplate amounts to $\alpha \approx 300–500$ cm^{-1} and is thus higher ($\alpha \geqslant 10$ cm^{-1}) compared to the wavelength of the Ho:YAG laser (λ = 2.1 μm) and lower ($\alpha \geqslant 2{,}000$ cm^{-1}) compared to that of the Er:YAG laser. Thus the absorption coefficients are higher for the Er:YAG and CO_2 lasers than for the Ho:YAG laser. Scholz and Grothues-Spork [12] attribute coupling of the erbium:YAG laser irradiation mainly to absorption to water and collagen and that of the CO_2 laser (λ = 10.6 μm) mainly to absorption by anorganic salts.

To assess the suitability of a pulsed laser for application in stapes surgery, it is a point of interest to determine the connection between the laser-specific parameters and the tissue effects at the footplate (ablation capacity, perforation, thermic marginal zones, etc.). This study does not examine other factors such as damage to surrounding structures, penetration of laser irradiation into the vestibule or heat, and noise production. However, they must be taken into account for a definitive statement about the suitability of these lasers for clinical use and will be subject of the subsequent publications.

MATERIALS AND METHODS

Forty isolated fresh human stapes and bovine compact-bone platelets (90 μm thick), which are comparable to the stapes in their laser absorption properties and perforation effects [2,4], were investigated to determine the effective laser parameters (wavelength, energy, beam diameter, repetition rate) and the suitable application technique for achieving a perforation diameter of 500–600 μm. The fresh preparations were kept refrigerated at −20°C until processing and irrigated with physiological saline immediately before laser application.

Of particular interest, apart from the attainable perforation diameters were the quality (shape and structure) of the perforations, the reproducibility of the perforation effect, and the thermically altered marginal zones occurring at the footplate. A total of 650 perforation diameters and the corresponding thermically altered mar-

ginal zones (crystallization, carbonization, and thermic transitional zones) were measured under a stereoscopic microscope (measuring accuracy of 10 μm) and brought in relation to the selected energy density. At each setting, five measurements were performed, and the mean value and variation range were determined. Histological (serial sections, HE staining) and scanning electron microscopic examinations of the stapes footplate provided additional information on the shape and structure of the perforations and the marginal zones.

Three pulsed laser systems were used: excimer (λ = 303 nm, pulse duration (FWHM) 0.12 μs, beam diameter 580 μm), Ho:YAG (λ = 2.1 μm, pulse duration (FWHM) 500 μs, beam diameter 550 μm), and Er:YSGG (λ = 2.78 μm, pulse duration (FWHM) 500 μs, beam diameter 480 μm) lasers. The laser beam was applied via a micromanipulator or fiber or in free emission.

RESULTS

Like the continuous-wave lasers, pulsed lasers also evidence a connection between the applied energy and the perforation diameter at the stapes footplate that manifests itself in three ranges of the perforating action: the nonperforating range, the threshold value with a rising phase and the saturation range of the perforation [2]. Taking into account these phases, the perforation effect and thermically altered marginal zones occurring at the footplate are found to differ relative to the applied laser systems and selected parameters.

Due to their distinct ablation mechanism, pulsed laser systems differ partially in their effect from the continuous-wave, thermically acting systems. An adequately large perforation can generally only be achieved by multiple shots at the same application site, since only a small amount of tissue is ablated per application. The requisite pulse count depends on the ablation rate of the applied laser and the selected energy density. An increase in the pulse count at the same energy density leads to a parallel shifting of the function to higher perforation diameters, as demonstrated using the Ho:YAG laser as an example (Fig. 1b). This results in more rapid attainment of the requisite perforation diameter at lower energy densities. To avoid applications of higher total energies, however, the energy densities must be chosen in such a way that the requisite perforation can be achieved with the smallest possible pulse count. We therefore define as effective an energy density range in which adequate perforation diameters can be achieved with the lowest energy densities. This lies at the beginning of the saturation range and is, for example, around 90 J/cm^2 for the Ho:YAG laser (Fig. 1b).

The attainable perforation diameters are also dependent on the diameter of the laser beam and, to a lesser extent, on the repetition rate. On irradiation of the same application site with low repetition rates ($\leqslant$10 Hz), the maximal perforation diameter corresponds to the laser-beam diameter in the range of the saturation phase. At higher repetition rates, the higher temperatures around the perforations due to inadequate cooling times between the pulses with a higher energy input in the marginal zones lead to additional bone removal and a slight enlargement of the perforation diameter. It is more effective, however, to enlarge the perforation diameter by increasing the beam diameter rather than the repetition rates. Moreover, lower repetition rates are associated with less extensive areas of thermically altered bordering tissue due to a low energy input in the marginal zones of the perforations, indicating an altogether lower thermic exposure.

Excimer Laser

The ablation rate per laser pulse of the used excimer laser is so low that a high pulse count (about 200) applied at the same site is needed to achive a maximal perforation diameter at the respective energy density. It can be perceived from the functional connection between the energy densities and the perforation diameters that, at a constant pulse count of about 200 and a pulse repetition rate of 2 Hz, the threshold energy density lies between 2.7 J/cm^2 and 3.3 J/cm^2 and, by a frequency increase to 10 Hz or 20 Hz, can be reduced to an energy density of between 1.9 and 2.3 J/cm^2 (Fig. 1a). The subsequent second range of the perforation action, the "rising phase", covers an energy density extending from about 2.3 J/cm^2 to about 5 J/cm^2. The perforation diameters vary in relation to the repetition rate and are greater at higher than at lower frequencies. The increase of function, on the other hand, is independent of the repetition rate.

The energy density at which the saturation range is reached is above 5 J/cm^2 irrespective of the repetition rate. The obtainable maximal perforation diameters, on the other hand, become

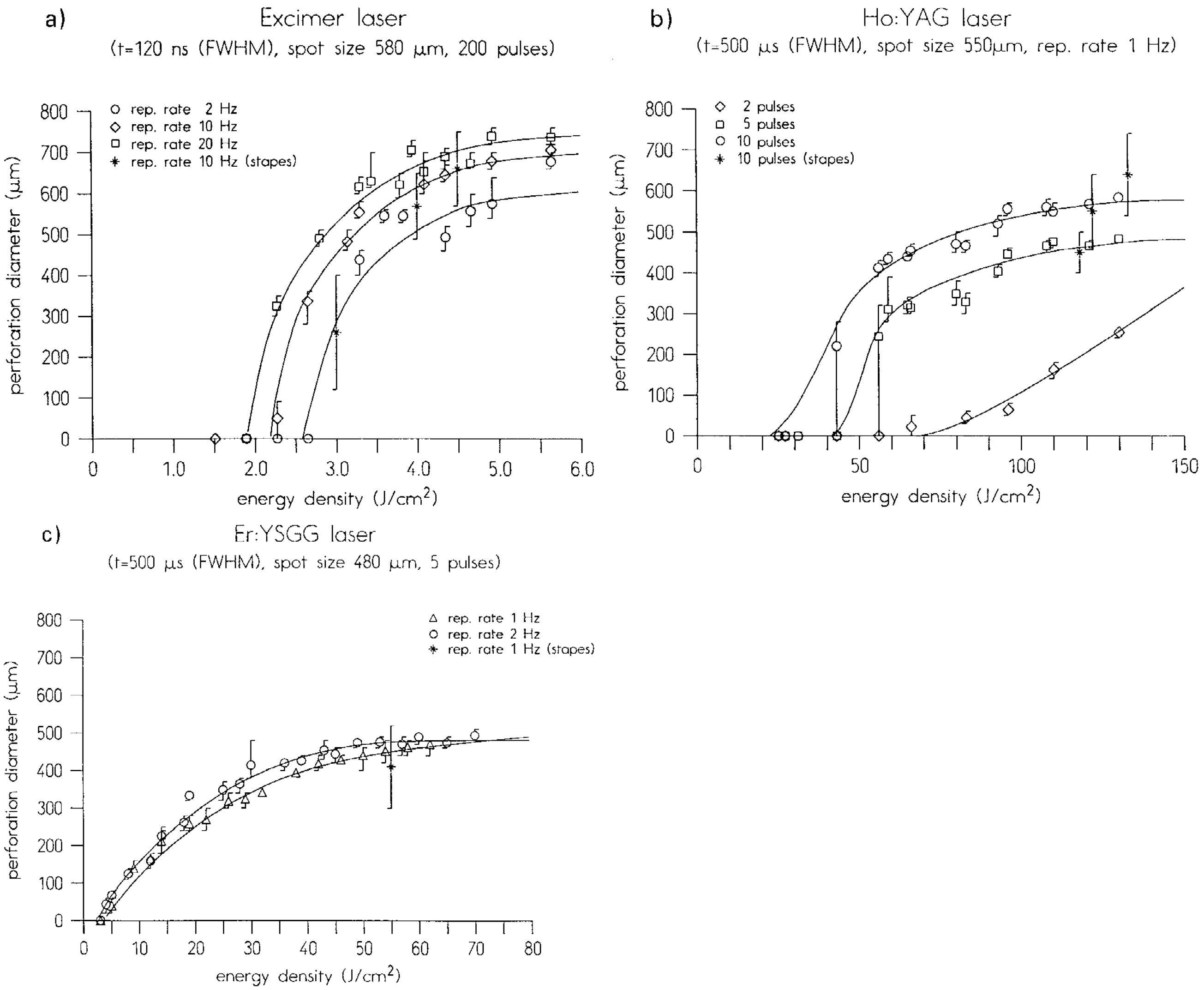

Fig. 1. Dependence of the perforation diameters on the energy density, repetition rate, and pulse count with the **(a)** excimer laser, **(b)** Ho:YAG laser, and **(c)** Er:YSGG laser.

greater with higher repetition rates at the same energy density. With a beam diameter of 580 µm and an energy density of about 5 J/cm², it is possible in the compact-bone model to achieve perforation diameters of about 560 µm at a pulse repetition rate of 2 Hz, about 680 µm at 10 Hz and about 740 µm at 20 Hz.

The perforation diameters determined in the human stapes footplate correspond to those achieved in the compact bone but exhibit a somewhat greater variation range (stapes: 180 µm; compact bone: 100 µm). This is due to the greater variation range of the examined footplate thicknesses (Fig. 1a).

At the perforation made with the excimer laser, there is an adjacent zone of reddish-brown discoloration (thermic transitional zone) that suggests thermic effects of the laser radiation (Fig. 2a). It varies in intensity depending on the repetition rate and the applied total energy. A frequency increase from 2 Hz to 10 Hz or 20 Hz results in a darker discoloration of this zone, which indicates higher thermic exposure of the tissue. Since crystallization and/or carbonization zones do not form with this system, even at higher repetition rates, an altogether low thermic tissue exposure can nevertheless be assumed, however.

This is also confirmed by histological examinations. The marginal zones of the perforation show only minimal thermic damage without carbonization. Scanning electron microscopy (Fig. 3a) reveals a somewhat rough perforation border

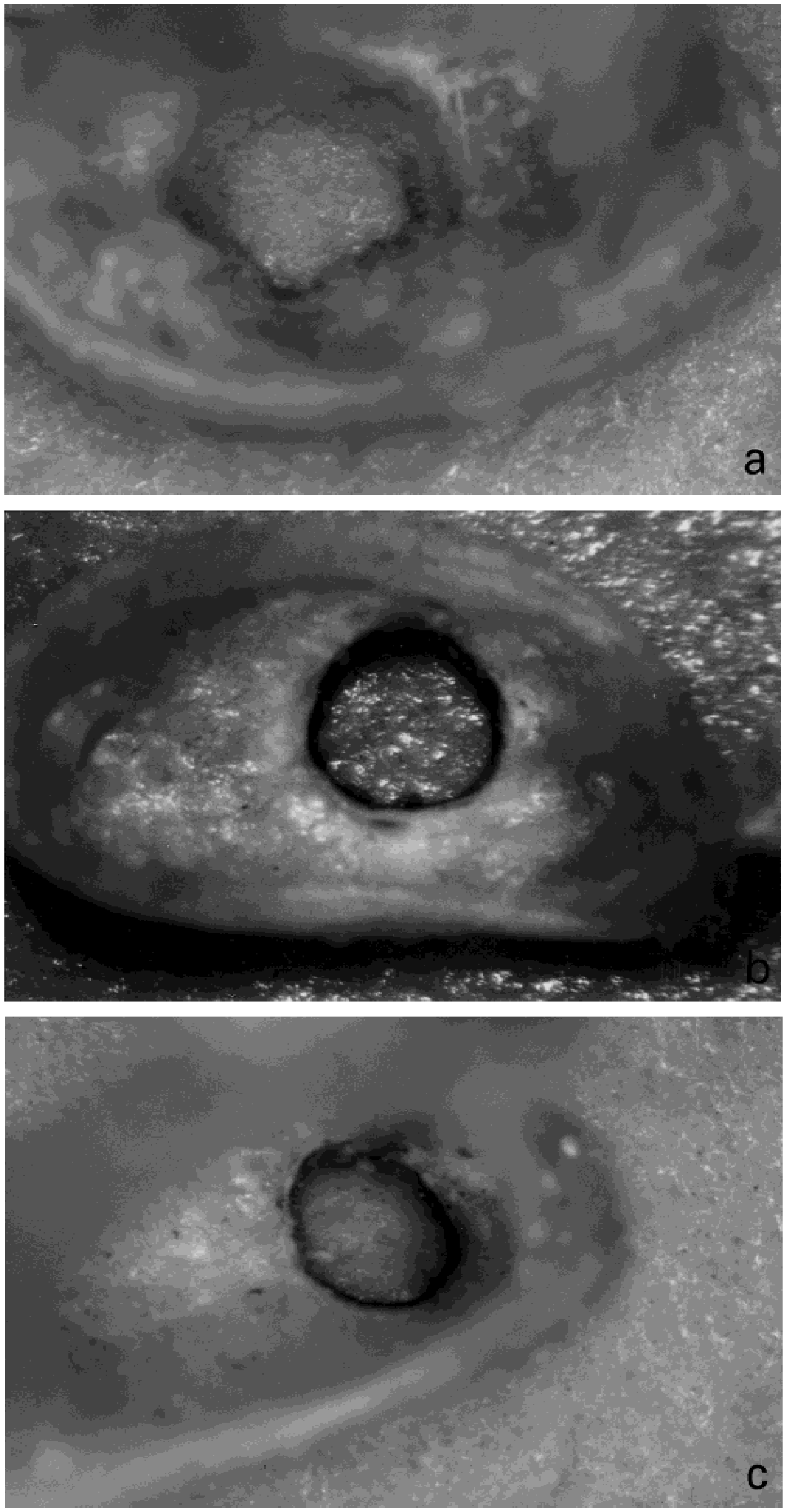

Fig. 2. Laser stapedotomy with the **(a)** excimer laser: energy density: 200 pulses at 5 J/cm^2 each; rep. rate: 10 Hz; diameter: perforation = 600–700 μm, no crystallization or carbonization zone, thermic transitional zone = 1000–1110 μm, **(b)** Ho:YAG laser: energy density: 10 pulses at 118 J/cm^2 each; rep. rate: 1 Hz; diameter: perforation = 540–560 μm, no crystallization zone, carbonization zone = 600–640 μm, thermic transitional zone = 680–720 μm, and **(c)** Er:YSGG laser: energy density: five pulses at 55 J/cm^2 each; rep. rate: 1 Hz; diameter: perforation = 390–500 μm, no crystallization zone, carbonization zone = 500–600 μm, thermic transitional zone = 550–620 μm.

with an adjacent, irregularly structured, wide thermic transitional zone without melting products as an indication of the action of low temperatures.

Effective parameters for achieving a perforation diameter of 500 μm to 600 μm were determined to be an energy density of about 5 J/cm^2, a repetition rate of 10 Hz and a pulse count of about 200 applications (Table 1).

Holmium:YAG Laser

Considering the low ablation power of this laser in the examined energy-density range, multiple irradiation of the same application site is necessary here too for perforation of the stapes footplate. About 10 pulses are required for a maximal perforation diameter at the respective energy density.

Figure 1b demonstrates the functional connection between the energy density and the attainable perforation diameters at a repetition rate of 1 Hz. The threshold value for a perforation with the Ho:YAG laser shows a wide range of variation and lies at energy densities of about 40 J/cm^2 for a repetition rate of 1 Hz. The rising phase of the function extends to an energy density of about 130 J/cm^2. With a laser-beam diameter of 550 μm, the maximally attainable perforation diameter is 580 μm for about 130 J/cm^2, a repetition rate of 1 Hz and 10 pulses. The necessary perforation diameters of 500–600 μm can be achieved with energy densities of 90 J/cm^2.

Application of the Ho:YAG laser in the human stapes with the determined effective parameters yields results similar to those in compact bone but with a considerably wider variation range. The mean perforation diameter at the footplate for 122 J/cm^2 is 550 μm (variation range 180 μm) as compared to 560 μm (variation range of 20 μm) in compact bone (Fig. 1b).

Just as with the excimer laser, use of the Ho:YAG laser at the footplate is associated with the occurence of thermically determined tissue alterations at the margin of the resultant perforations (Fig. 2b). A narrow carbonization zone forms in addition to the thermic transitional zone already observed with the excimer laser. The perforation is nearly round and has a smooth contour.

The histological picture of a footplate treated with the Ho:YAG laser discloses a marginal zone of the perforation evidencing only minimal thermic damage with an otherwise intact bone structure. Scanning electron microscopy of a footplate after Ho:YAG laser application reveals an evenly

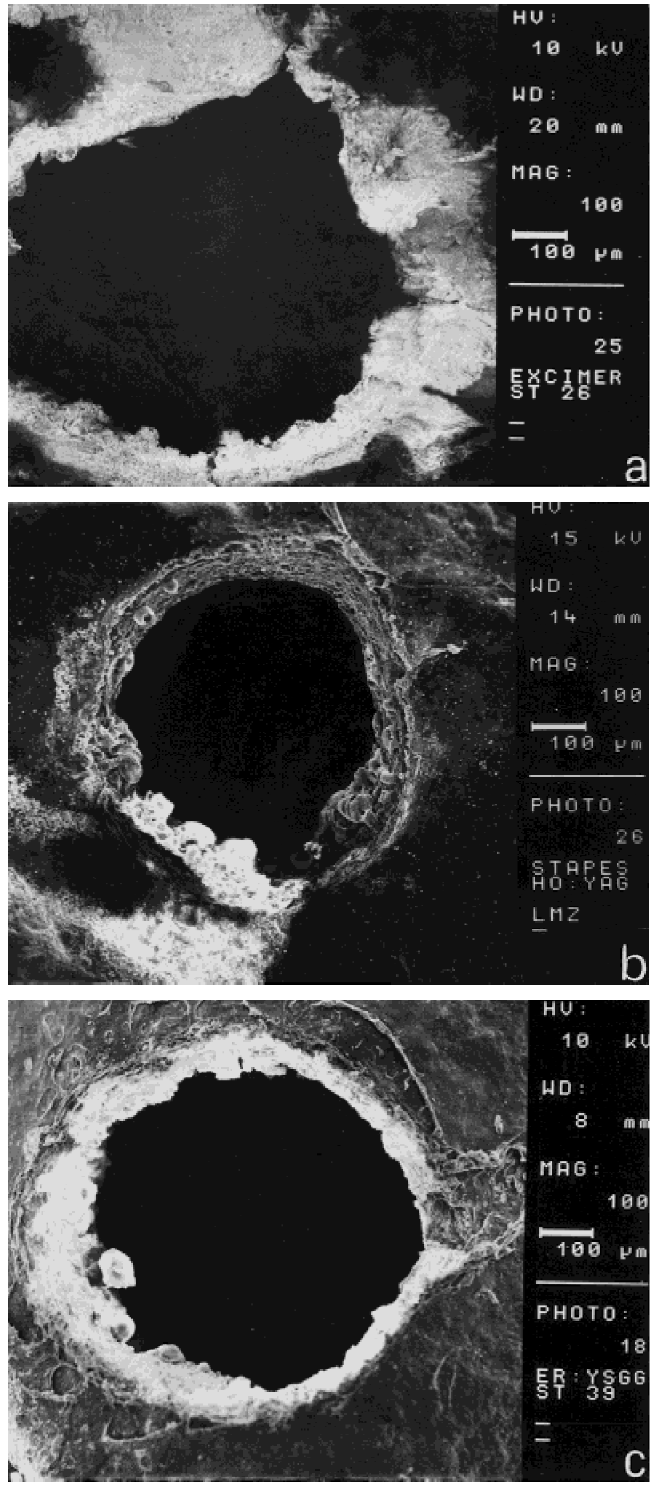

Fig. 3. Scanning electron microscopic image of a footplate perforation with the **(a)** excimer laser, **(b)** Ho:YAG laser, and **(c)** Er:YSGG laser.

TABLE 1. Effective Laser Parameters for a Stapes Footplate Perforation With a Diameter of 500–600 μm, Taking Into Account the Total Energy Required and the Laser Effect in the Tissue*

Laser system	Beam diameter	Energy density	Rep. rate	Number of pulses	Application duration	Total energy	Perforation quality	Reproducibility	Low thermic side effects
Er:YSGG 2.78 μm, 500 μs	≈550 μm	55 J/cm^2	1 Hz	≈5	5 s	≈0.5 J	+++	+++	++
Ho:YAG 2.1 μm; 500 μs	≈550 μm	90 J/cm^2	2 Hz	≈10	5 s	≈2.1 J	+++	+++	++
Excimer 308 nm; 120 ns	≈580 μm	5 J/cm^2	10 Hz	≈200	20 s	≈2.6 J	++	+	+++

*+++, Very good; ++ good; + satisfactory; − poor.

rounded perforation with regular margin configuration and a narrow irregular carbonization zone (Fig. 3b).

Effective parameters for a 500–600 μm perforation diameter were determined to be an energy density of about 90 J/cm^2, a repetition rate of 1 Hz and a pulse count of about 10 applications (Table 1).

Erbium:YSGG Laser

Here too, perforation of the stapes footplate can only be achieved by multiple irradiation of the same application site. Higher ablation rates, however, lead to a lower pulse count of about five pulses, which are necessary for a maximal perforation diameter at the respective energy density.

Figure 1c demonstrates the functional connection between the energy density and the obtainable perforation diameters for various repetition rates. The threshold value at which a compact-bone perforation is first achieved varies negligibly in relation to the applied repetition rate (1 Hz and 2 Hz) and is below 5 J/cm^2. The rising phase extends over an energy-density range of about 50 J/cm^2 and then passes into the saturation phase. The increase in function is not altered appreciably by raising the repetition rate to 2 Hz. A maximal perforation diameter of about 480 μm is achieved with a beam diameter of about 480 μm. Considering the dependence of the maximally obtainable perforation diameter on the applied laser-beam diameter, beam diameters of about 550 μm and energy densities of over 50 J/cm^2 are necessary with the Er:YSGG laser to achieve perforation diameters of 500–600 μm.

The mean perforation diameter of 410 μm obtained at the stapes footplate with a beam diameter of 480 μm is somewhat smaller than that in the compact-bone model and shows a greater variation range (Fig. 1c).

The treatment of footplates in both the Ho:YAG laser and Er:YSGG laser results in the formation of a uniform, nearly round perforation with only slightly evident thermically altered marginal zones (Fig. 2c). A narrow carbonization zone and a thermic transitional zone are formed. They are already detectable at low pulse repetition rates. The histological picture of a footplate treated with the Er:YSGG laser shows slight thermic bone damage in the perforation area with a narrow carbonization border (Fig. 4). Scanning electron microscopy of a footplate after application of the Er:YSGG laser discloses an evenly rounded perforation with a somewhat irregular margin and a barely discernible carbonization zone (Fig. 3c).

Effective parameters for a 500–600 μm perforation diameter were determined to be an energy density of about 55 J/cm^2, a repetition rate of 1 Hz and a pulse count of about five applications (Table 1).

DISCUSSION

It was the primary objective of this study to determine the laser parameters which permit effective stapes management with a suitable application technique and which are necessary for achieving a footplate perforation with a diameter of 500–600 μm.

Compared to radiation of the excimer (λ = 308 nm) and the Ho:YAG (λ = 2.1 μm) wavelength, that of the Er:YSGG wavelength (λ = 2.78 μm) is more strongly absorbed by bone tissue. The Er:YSGG laser has about 30 times higher and the Ho:YAG laser about 20 times higher ablation rates than the excimer laser sys-

Fig. 4. Flat section through the stapes footplate after Er:YSGG laser perforation (serial section with a 4 m slice thickness, HE staining, 10-fold).

tem. Therefore, the Er:YSGG laser requires not only a lower pulse count (about five pulses) but also less energy to achieve an adequately large perforation. The somewhat lower ablation rate of the Ho:YAG laser leads to a pulse count of about 10 pulses and an approximately four times higher total energy (Table 1).

Of the pulsed laser systems examined, the Er:YSGG laser thus evidences the highest ablation rate at a footplate and is therefore the most effective laser for stapedotomy. This is also confirmed by studies with the Er:YAG laser (λ = 2.94 μm), whose absorption properties and thus footplate effects are comparable to those with the Er:YSGG laser [10,11]. Though somewhat less effective than the Er:YSGG laser according to our results, the Ho:YAG laser likewise appears to be well suited for perforation of the footplate [7,14].

The quality of the perforations, the reproducibility of the perforation diameters and the extent of thermic side effects at the footplate are comparable with the Er:YSGG and the Ho:YAG laser. Both laser systems create round perforations with a regularly shaped border and only sligthly evident thermic marginal zones.

The applied excimer laser (λ = 308 nm, pulse duration 120 ns) does not appear to us to be well-suited for stapedotomy in view of its low ablation rate at the footplate. The high pulse count of about 200 single pulses leads to a long application time of about 20 seconds even at a relatively high repetition rate of 10 Hz and could cause technical difficulties due to the necessity of keeping to the same application site during the entire irradiation period. Its use thus appears to be problematic from a clinical point of view. Moreover, at approximately 2.6 J, the total energy required is two and a half times higher than with the Er:YSGG laser. Nevertheless, the marginal zones of the perforations show less thermic damage than with any of the other laser systems applied, which is morphologically manifested in the absence of a carbonization zone. This is due to the low absorption of the UV wavelength of this laser at the footplate and to its extremely short pulse duration (120 ns), which leads to a lower energy input in the marginal zones and thus to a lower degree of heating. A multiple laser application with pulse repetition rates of over 10 Hz, however, causes a broadening of the thermic marginal zones, so that the use of higher repetition rates do not appear useful despite the requisite high pulse count of 200 pulses or more. Compared to the other lasers, perforation with the excimer laser is of lower quality (irregular perforation border), and the reproducibility of the perforation diam-

eters is poor due to strong scattering of the ultraviolet radiation in the tissue and thus considerable variation in the ablation capacity.

The use of other wavelengths in the UV range and shorter pulse durations could, however, improve the ablation capacity of the excimer laser [12,13]. Segas et al. [13], for example, applied an excimer laser with a wavelength of 193 nm and a pulse duration of 15 ns and needed a pulse count of 20 to 125 pulses for perforation of the footplate.

The tissue-ablating effect of pulsed laser systems, photoablation, permits precise and controlled stapes footplate perforation through low and readily reproducible ablation rates. The extent of thermic side effects at the footplate is lower in comparison to the purely thermically acting cw and superpulse laser systems [4]. The higher pulse count needed for a perforation, however, could prove to be disadvantageous in clinical practice owing to the longer application time and the necessity of irradiating the same application site. Furthermore, because of the marked variation in the thickness of most stapes footplates, pulsed lasers with multiple applications to the same site will not achieve a smooth perforation without the irradiation penetrating into the inner ear. This fact and the laser-induced shock waves in the perilymph of these photoablative systems suggest that these lasers may be more dangerous for the inner ear than cw lasers. This has to be clarified in subsequent studies.

Considering the requisite total energy and the quality of the perforations, the pulsed lasers have distinct advantages over the continuous wave systems. The total energy needed for an adequately large perforation is lower by the factor 2–4 with the Er:YSGG laser than with the CO_2 laser in cw and the superpulse mode. Similarly high total energies are required with the excimer as with the argon laser, which is an indication of the lower effectivity of these lasers. Though an effective instrument for stapedotomy, the Ho:YAG laser likewise leads to high total energies.

The quality (shape and structure) of perforations is generally better with pulsed than with cw lasers [4], although well-shaped uniformly round perforations can also be made, for example, with the CO_2 laser using the superpulse mode [6] or the application technique with microprocessor-controlled rotating mirrors in cw mode [5].

Based on the experimental results presented regarding the tissue effect of the examined pulsed laser systems, a statement can be made about their suitability for stapedotomy. Accordingly, the Er:YSGG and the Ho:YAG lasers have been found to be suitable systems for the fenestration of the stapes footplate. On the other hand, the applied excimer laser system must be regarded as unsuitable (Table 1). Thus the erbium laser, which has proven to be the most effective system from the group of the investigated pulsed lasers, could represent an alternative to the argon, KTP 532, and CO_2 lasers, already successful when clinically applied in stapes surgery [3,6,9,15].

To make a definitive statement about the safest and most effective laser system for stapes surgery, however, it is of the utmost importance to consider as suitability criteria, not only the ablation rate at the footplate and suprastructures, but also the transmission of thermic energy into the vestibule and the acoustic stress to the inner ear during laser stapedotomy. These phenomena have already been investigated and will be reported in subsequent studies.

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